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Abstract

The knowledge of the dose and of the energy spectrum of the X-rays delivered to the patient during a radiological examination allows in principle the computation of the number of photons per unit surface useful for a good mammography.

The film-screen assembly detectors used in the present standard practice require a number of photons per unit surface which, from a statistical point of view, would be unnecessarily high if single photon counting detectors with efficiency near to one were available.

We discuss a possible solid state detector with these characteristics. Moreover, we propose the use of an X-ray monochromatic beam from a synchrotron radiation source in order to perform the examination at the energy where the signal to noise ratio has a maximum.

Using the proposed detector in such a beam a substantial dose reduction can be foreseen.

Introduction

The basic unit of information in an imaging system is the detected photon. An image corresponds to the knowledge of the number of photons impinging on some photosensitive device at specified locations. This knowledge allows one to quantitatively define the quality of an image. In the common radiological practice the dose and the energy spectrum of the X-rays delivered to the patient are known. This in principle allows one to compute the number of photons per unit surface which must be present in order to obtain a good radiological examination. When examining soft tissues, in mammography for instance, soft X-rays (~20 keV) are generally employed and the typical dose is around 1.7 mGy, corresponding roughly, at that energy, to 10^7 photons/mm^2 impinging on the surface of the sample. Consider now a sample to be imaged by means of a detector the surface of which is subdivided in pixels. Given a typical absorption of 60%, around the energy of interest, this would result in an incident planar fluence [1] on the surface of the detector of $4 \cdot 10^6$ photons/mm^2. Let us now define the following quantities: n_b is the number of photons incident on a pixel looking at the "background", n_d is the number of photons incident on a pixel containing some detail to be imaged which we assume has absorbed a percentage p of the original incident photons; it is $n_d = (1-p)n_b$. We can then define the *signal* as $n_b - n_d = pn_b$, and the *noise* as $\sqrt{n_b + n_d} = \sqrt{n_b(2-p)}$. The *contrast* is given by

$$\frac{n_b - n_d}{\frac{n_b + n_d}{2}} = \frac{2p}{2-p} = c \quad , \qquad (1)$$

while the *signal-to noise ratio* can be written as

$$k = \frac{pn_b}{\sqrt{n_b(2-p)}}. \qquad (2)$$

It follows that the number of "background" photons needed to "see" a detail with a given contrast c is $n_b = 2k^2/pc$. If $p < 0.2$ then $c \approx p$, and one writes

$$n_b \approx 2\left(\frac{k}{c}\right)^2. \qquad (3)$$

Figure 1 shows a plot of n_b versus contrast for a few values of k. The signal-to-noise ratio must be chosen high enough so that, in a given set of pixels, the probability of any pixel giving accidentally a false signal is negligible. For 10^5-10^6 pixels this condition is met by setting $k = 5$ [2].

Assume now one has a detector with an active surface A subdivided in square pixels of side d. If N total photons are incident on the surface A, then $N/n_b = A/d^2$, and from Eq. 3 it follows that

$$c = \frac{\sqrt{2}k}{d\sqrt{N/A}}. \tag{4}$$

Inserting in Eq. 4 above the numerical values $N/A = 4{\cdot}10^6$ photons/mm^2, d = 0.2 mm, and k = 5 one finds c = $1.7{\cdot}10^{-2}$. This contrast is extremely low, and in fact the practical rule is that the minimum contrast perceivable by the human eye is ~2.5%. From a statistical point of view then, the number of photons per unit surface employed in a conventional radiological examination would appear to be excessive. A detector with near unity efficiency and capable of single-photon counting would require less photons, and therefore a lower radiation dose. An optimal choice of the radiation source used to illuminate the sample also contributes to the goal of improving the quality of radiological images while reducing the radiation dose. The signal-to-noise ratio, in the case of soft tissues, is a function of energy, and MonteCarlo calculations [3] show that a maximum is found at about 20-25 keV. A monochromatic source would then allow one to select the range of energies where the signal-to-noise ratio is at a maximum, and to avoid the additional radiation dose delivered for example by a conventional X-ray tube. The SYRMEP (SYnchrotron Radiation for MEdical Physics) [4,5] project addresses both the source and the detection problems. A monochromatic X-ray beam with energies between 15 and 30 keV will be provided by a bending-magnet beamline, devoted to medical physics, presently under construction at the Elettra synchrotron light source in Trieste, Italy. At a distance of 20 m from the bending magnet the laminar beam will have a 4$\times$150 mm^2 section and will illuminate a pixel, high efficiency, silicon detector having the same cross-section. In this way the noise due to scattering will be strongly suppressed eliminating the need for an anti-scatter grid. A prototype of this detector with its associated electronics has been successfully built and tested [6].

Imaging with a digital detector

As evidenced above, a quantum of information is obtained from each detected photon. Photon production is a poissonian process, and therefore the number of detected photons will be distributed according to Poisson statistics. Then, if in a unit time and per unit area one counts an average number $\bar{n}$ of photons, the standard deviation will be $\sqrt{\bar{n}}$. This constitutes a fluctuation in the single measurement which is called quantum noise. In the sole presence of quantum noise one speaks of a quantum limited image. This noise is intrinsic to the mechanism of photon counting and is therefore unavoidable. If we take into account only the quantum noise and we assume that our detector has a 100% efficiency, then Eq. 4 holds when the smallest detail to be imaged has the dimension of a pixel. Eq. 4 also shows that contrast varies inversely with the square root of the superficial dose (which is proportional to the number of photons per unit surface impinging on the detector). When the dimensions of the object to be imaged are different from the pixel size, Eq. 4 must be replaced by a more involved expression having the form

$$C_m = \frac{1}{\sqrt{N/A}} \frac{\mathrm{k}(a)}{\sqrt{a}}, \tag{5}$$

where k(a) is a complicated function of the object area a [7]. From a qualitative point of view, one can say that if a "small" object is to be detected, then it must have a large contrast, while "large" objects may be imaged even with a small contrast. C_m is the minimum contrast which can be detected using a given number of photons per unit area and which could be ideally achieved with a 100% efficient noiseless detector. A real detector with its electronic chain, however, will have an efficiency less that 100%, thereby reducing N/A, and will also add noise to the detected photon counts. If one then calls R the ratio between the total noise, including the detector noise, and the intrinsic quantum noise, then the minimum detectable contrast becomes

$$C = \frac{R}{\sqrt{\varepsilon}} C_m, \tag{6}$$

where ε is the detector efficiency. Eq. 6 above shows that a poor efficiency yields a minimum detectable contrast higher than the ideal case C_m, degrading therefore the performance of the detection system, and the same does a noisy detector.

Test object called phantoms are routinely used in mammography to test the performance of imaging systems. A phantom made of a 6 mm diameter, 75 µm thick, aluminium disk embedded in plexiglas can be readily imaged using a conventional X-ray tube and a film as the detector. This is to be expected, since the fluence is quite high as discussed above. On the other hand, when employing a digital detector the situation is radically different. Figure 2 shows several images of the above detail obtained by means of the SYRMEP silicon detector [6] illuminated with increasing radiation fluences. These images clearly evidence that the object is already visible at a fluence of

$6.9 \cdot 10^3$ photons/mm^2 which is a factor of ~10^3 lower than in the case of a film plate detector. Also, it can be seen that an increase in fluence corresponds to an increase in image contrast, and therefore to an increase in object visibility. This is correctly predicted by Eq. 4 above. A digital detector has a further advantage: once the image has been formed as a set of numerical data, the depth of the grey-scale can be changed at will to enhance contrast or, better, greys can be substituted, as shown in Figure 2, by false colours.

Spatial resolution and image reconstruction

In the previuos discussion, the size of the smallest detail which could be detected, and therefore the spatial resolution of the detector, was assumed to coincide with the pixel size. Spatial resolution however, need not be limited by pixel size. Images of phantoms detailing a spatial resolution lower that pixel size have already been produced using the SYRMEP detector. The sample is imaged by scanning in successive steps the detector across the whole sample surface by means of a micrometric movement stage. The spatial resolution which can be achieved in the scanning direction is then given by the width of the scanning step. In principle then, the spatial resolution could be made arbitrarily small [8], at least in one direction. This however carries the price of an increased radiation dose, since the sample must be irradiated by a longer total time. Raw data obtained in a scanning run must be processed to yield the final image. Image processing and reconstruction are briefly discussed in Ref. 6 and references therein.

References and notes

[1] For a definition of planar fluence see for instance F. Attix and W.C. Roesch, Radiation Dosimetry, Second Edition, Vol 1, p. 21-22, Academic Press, New York (1968).

[2] A. Ll. Evans, The Evaluation of Medical Images, Medical Physics Handbooks 10, Adam Hilger, Bristol (1981).

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[4] F. Arfelli et al., SYnchrotron Radiation for MEdical Physics. A comparison between digital and conventional screen-film images, Phys. Med., Vol. IX (1993) p.175.

[5] F. Arfelli et al., SYRMEP(SYnchrotron Radiation for MEdical Physics). Performance of the digital detection system, Phys. Med., Vol. IX, Suppl. 1 (1993) p.229.

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[7] M. Di Michiel, Un rivelatore di silicio a pixel per immagini in radiologia diagnostica, Thesis, Università di Trieste (1994), unpublished.

[8] Current, commercially available, micrometric movement systems can achieve sub-micron positioning accuracy.

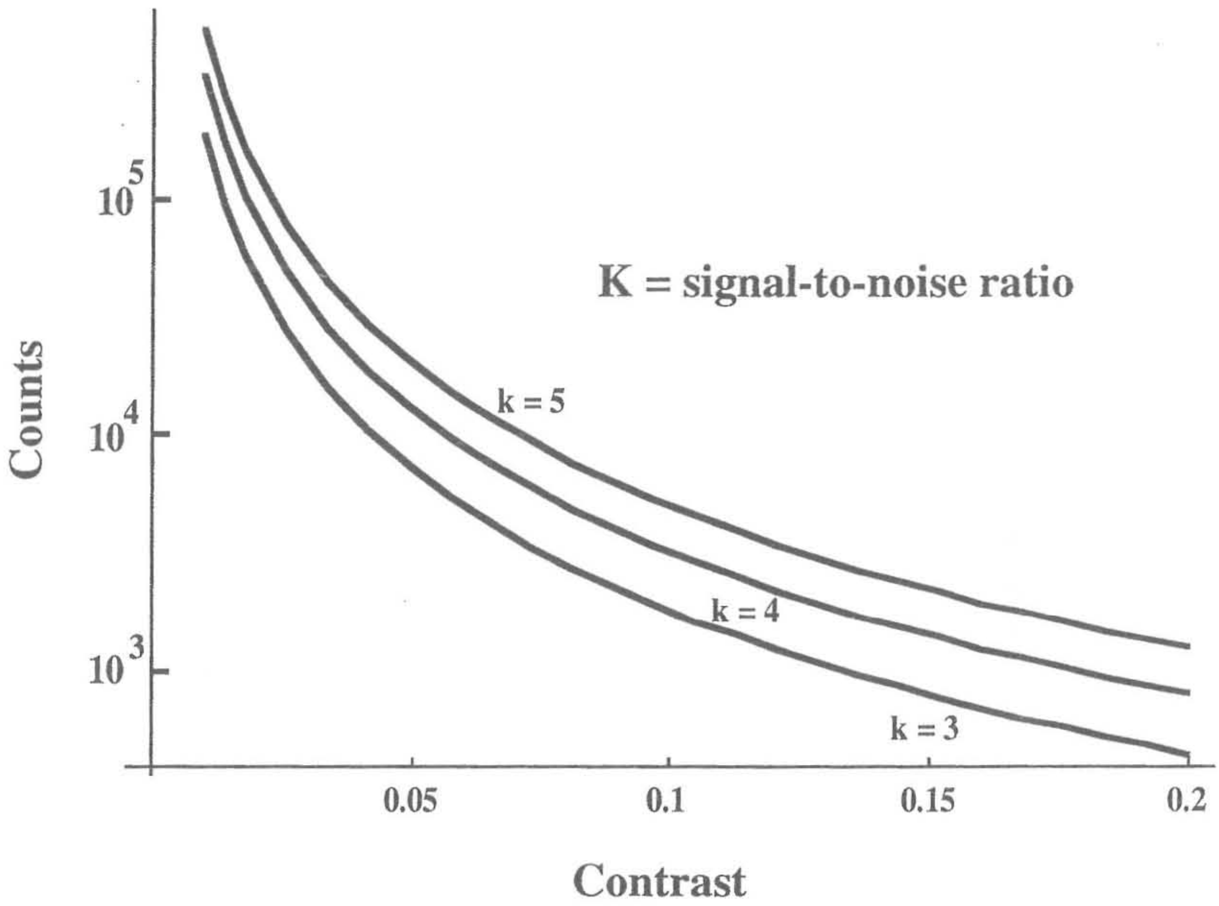

Figure 1: Visibility curves (see text).

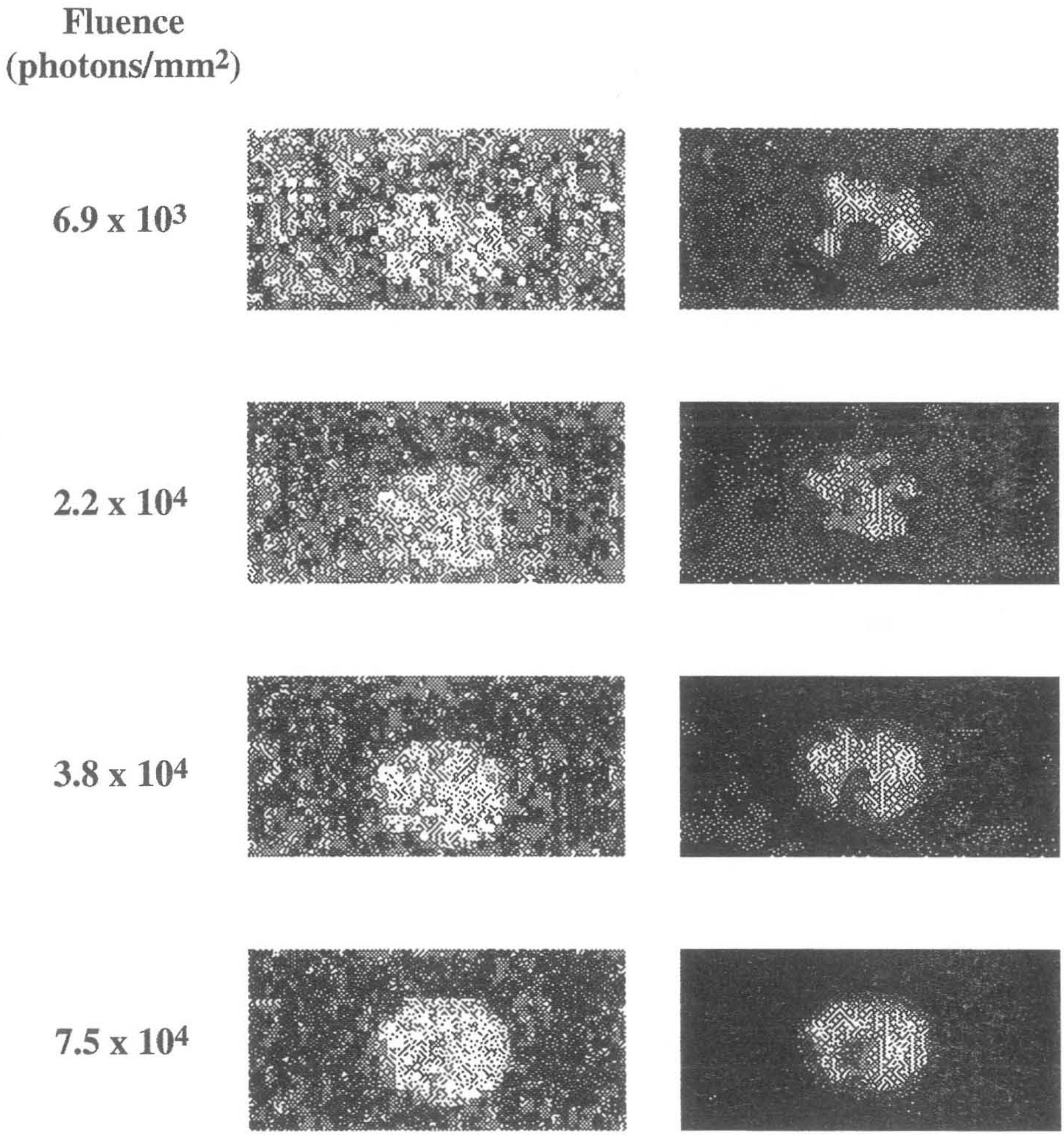

Figure 2: Digital images of a detail from a standard mammographic phantom. The detail is a 6 mm diameter, 75 μm thick, aluminium disk embedded in plexiglas. Images are shown in order of increasing fluence on the detector. False colour renditions are shown on the right hand side column.